Automated detection of soleus concentric contraction in variable gait conditions for improved exosuit control

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***Abstract*—Exosuits can reduce metabolic demand and improve gait. Controllers explicitly derived from biological mechanisms that reflect the user's joint or muscle dynamics should in theory allow for individualized assistance and enable adaptation to changing gait. With the goal of developing an exosuit control strategy based on muscle power, we present an approach for estimating, at real time rates, when the soleus muscle begins to generate positive power. A low-profile ultrasound system recorded B-mode images of the soleus in walking individuals. An automated routine using optical flow segmented the data to a normalized gait cycle and estimated the onset of concentric contraction at real-time rates (~130Hz). Segmentation error was within 1% of the gait cycle compared to using ground reaction forces. Estimation of onset of concentric contraction had a high correlation (R^2=0.92) and an RMSE of 2.6% gait cycle relative to manual estimation. We demonstrated the ability to estimate the onset of concentric contraction during fixed speed walking in healthy individuals that ranged from 39.3% to 45.8% of the gait cycle and feasibility in two persons post-stroke walking at comfortable walking speed. We also showed the ability to measure a shift in onset timing to 7% earlier when the biological system adapts from level to incline walking. Finally, we provided an initial evaluation for how the onset of concentric contraction might be used to inform exosuit control in level and incline walking.**

I. INTRODUCTION

The complexity of the human-robot system makes the development of assistance strategies that account for the variability in individual gait patterns difficult. As an example, reductions in metabolic cost with exoskeleton or exosuit assistance vary widely across individuals, and assistance parameters, such as the time of applied force, that work well for some individuals are ineffective or detrimental for others [1]. Although we have a good understanding on the robotic side of the human-robot system, we still lack a good understanding of how individuals interact effectively with such devices. Due to difficulty in linking human mechanics to effective control strategies, much work has focused on determining parameters via a grid sweep or optimization [2, 3]. However, these approaches are time-consuming, determined for one gait condition, and are not necessarily generalizable or adaptive to differences in gait and terrain.

An approach that is derived from the biological mechanics of the individual should, in theory, allow the device to interact intuitively with the individual and adapt to user changes associated with gait (*e.g.* incline, speed). Indeed, wearable robotic systems have been developed that take this approach either through matching the muscle activation pattern (*i.e.* electromyography (EMG)) or by matching the kinematics or kinetics of the ankle joint [4-6]. However, these approaches often do not achieve performance on par with sweeps or optimization [2, 3]. We suspect that this is because EMG and ankle joint kinematics, torque and power do not completely capture the dynamic state of ankle plantarflexor muscles. Towards the larger aim of providing customized bio-inspired assistance that maximizes metabolic improvement, we posit that assistance from the exosuit should reflect the power of the muscle and should thus assist when the muscle is concentrically contracting. For our ankle exosuit design which assists with plantarflexion power at push off, we hypothesize that applying assistance coincident with concentric contraction of the largest calf plantarflexor, the soleus, will lead to improved metabolic reduction across gait conditions relative to a standard timing.

To explore this, we need the ability to estimate muscle-tendon kinematic behavior in real-time during walking. The key feature missing from more common biological measurements (joint kinematics or kinetics, power, or EMG), that would allow us to make a calculation of muscle power is an understanding of whether the muscle is shortening (concentric contraction), static (isometric contraction), or lengthening (eccentric contraction). Due to the compliant Achilles tendon (AT) in series with the ankle plantarflexor muscles, the kinematics of the plantarflexor muscles cannot be accurately determined from the joint kinematics alone [7]. EMG is an estimate of muscle activation and predicting muscle dynamics from EMG requires many assumptions and an understanding of whole-body dynamics [8-10].

The common approach for measuring muscle kinematics of the triceps surae in humans is B-mode ultrasound (US) imaging [11-13]. This approach provides detail about the absolute length, angle, and velocity of the muscle fascicles, but the post-processing time for previous approaches, including semi-automated routines, can be substantial and manual corrections are often required [11, 14]. In previous studies, these data are then averaged across individuals, rendering individual variability that could inform

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development of tailored controllers lost. This image capturing and post-processing requirement and subsequent lack of individuality limits the use of muscle dynamic behavior for human-in-the-loop tailored assistance profiles. While our approach provides less information about muscle state compared approaches that require post-processing or multiple probes, the onset of muscle concentric contraction may provide enough information for exosuit control.

In this work, we present an approach that uses US to rapidly detect when the muscle begins to concentrically contract (*i.e.* generates positive power) just before ankle push-off. We then show how the technique can be used to measure the individuality of contraction timing (healthy and poststroke) and the adaptation of contraction timing to changing gait (*e.g.* incline). Finally, we provide an initial example of how this might be used for improving exoskeleton and exosuit control for assisting in level and incline gait.

The paper is divided into (II) an automated muscle tracking technique capable of gait segmentation and estimation of onset of concentric contraction at real time rates in walking, (III) validation of the tracking algorithm, (IV) use of the approach for evaluating the variability in muscle contraction patterns across individuals (including persons post-stroke) and across changes in incline grade, and (V) initial evaluation of how onset of muscle concentric contraction can inform exosuit control in level and incline walking.

II. Muscle Tracking Algorithm

A. Biomechanical Muscle Mechanics in Walking

Prior studies suggest that for the first ~40% of the gait cycle, the muscle contraction state is ‘almost isometric’ [15] where the tendon can stretch against the muscle and store energy while the muscle is producing zero mechanical power [13]. During the late phase of stance, the muscle concentrically contracts (shortens under load) and actively produces positive power [13]. The goal of our approach is to detect when the muscle is concentrically contracting just before push-off.

We modeled the ankle soleus-AT complex as an elastic element and contractile element in series (Fig 1). We tracked the displacement of the superficial soleus to determine whether the muscle was displacing proximally (away from the ankle) and shortening against the distal AT. We chose the soleus because the soleus is the largest calf muscle and may have a more important role in walking [15]. Additionally, the soleus muscle dynamics are unaffected by knee kinematics [16].

This approach should be less susceptible to noise and drift because we are looking at changes in muscle velocity rather than absolute lengths. Although ankle plantarflexion can lead to aponeurosis displacement [17], because the ankle is dorsiflexing during the time period of interest, proximal tissue displacement can be associated with concentric contraction. Finally, despite inability to determine the length of the muscle fascicle, we can determine whether the muscle is displacing against the distal AT, contributing positive AT strain, and beginning to decelerate the ankle just before transition to ankle plantarflexion [17] (Fig 1).

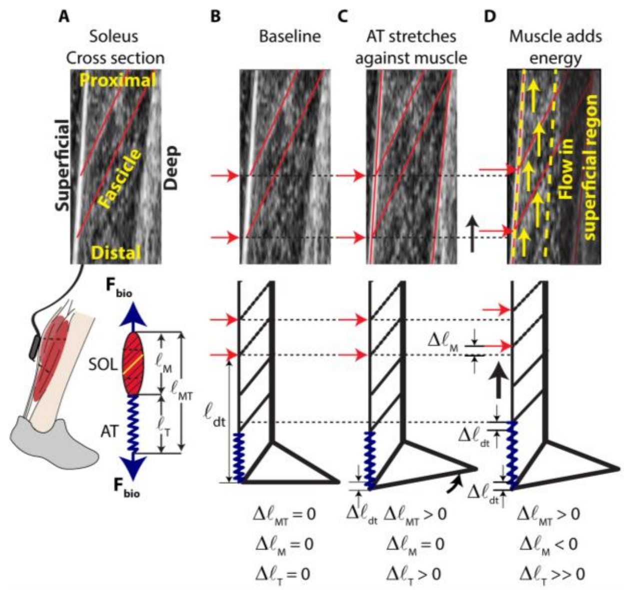

Figure 1: Ultrasound images of triceps surae and cartoon depictions of muscle-tendon interactions. (A) US probe was placed over the soleus (SOL). The muscle-tendon unit (MTU) was modeled as a muscle and tendon in series. (B) We tracked the displacement of the soleus fascicle at the insertion into the superficial aponeurosis. Relative displacement was associated with movement of the distal Achilles tendon (AT). (C) During dorsiflexion, the ankle rotates such that the MTU gets longer while the muscle maintains an isometric contraction which causes the tendon to stretch. (D) Just before plantarflexion, the MTU is still lengthening. In addition, the muscle begins to shorten and adds additional energy.

B. Ultrasound Image Capture

We securely attached a low-profile US transducer (MicrUs, Telemed, Lithuania) over the calf muscle using self-adhesive wrap tight enough to prevent movement against the skin [18]. With the subject walking in various conditions, we captured B-mode images of the soleus muscle at ~113 Hz. The US system exported a pulse at each frame which we used to sync with the motion capture (Qualisys, Sweden) and ground reaction force (GRF) data (Bertec, USA). The US data was batch-captured and then the stack of images was tracked. For each condition, we captured approximately 1300 frames, which equates to ~10 steps or 10 seconds.

C. Muscle - Tendon Tracking

Once the US images were captured, a custom MATLAB (MathWorks, USA) algorithm extracted the flow data on an Intel Core i7-8750H 2.2Ghz processor (Fig 2). The algorithm processed the US data at real time rates (~130 Hz) which were greater than the frequency of the US system (~113 Hz). We first manually initialized the automated algorithm by identifying the region of interest (ROI) (Fig 2A). For this work, the ROI was the superficial half of the soleus adjacent to the superficial aponeurosis. We used a small ROI of approximately 15 x 5 mm (vertical length x depth) which allowed for high processing rate while maintaining accuracy. A sparse-to-dense optical flow subroutine (Fig 2B) (MATLAB, OpenCV) tracked the image flow, and the flow vector was rotated such that only the component of the vector that was parallel to the aponeurosis was extracted. The muscle was calculated to be concentrically contracting when the flow was directed proximally, away from the ankle.

Because US is needed to detect the muscle dynamic behavior, it can also be used for gait segmentation in overground settings, thus using only a single sensor and avoiding sensor fusion. Accurate segmentation is important as it enables the data to be captured as a function of a normalized gait cycle, and thus allows for averaging of data across multiple strides. This can be used to aid in analysis of the data or to assist with control of an exoskeleton or exosuit. To accomplish this, the gait was segmented from heel-strike to heel-strike using the flow data (Fig 2B). Due to rapid displacement of the muscle due to foot contact and rotation of the ankle, we observed that a rapid transition from negative to positive flow (zero crossing) occurs at heel-strike (foot-contact). The accuracy of this approach was verified against the GRF and is detailed in section IIIA. The segmentation subroutine (Fig 2C) identified these large transients to determine when heel-strike occurred. The flow was normalized to 1000 data points and the strides were averaged to obtain an average stride (Fig 2C).

To calculate onset of muscle concentric contraction (*i.e.* to discriminate the transition from 'almost isometric' to shortening), the threshold subroutine calculated the first time when the when the flow velocity was (1) positive above a manually determined threshold (0.0002) and (2) the flow velocity continued to increase (Fig 2D). For (2), we applied a moving 50 datapoint (5% GC) linear fit across the time region and determined the index in which the slope of the linear fit was greater than a manual threshold (0.02). Within the region of 30% to 55% of the GC, the onset was determined as the first time both requirements were met. This method was based on the fact that the muscle should be concentrically contracting continuously towards rapid push-off. The same values for the threshold were used for all trials and participants.

Two major advantages of our approach are the ability to work on a minimized US field of view (FOV) and the ability to perform the calculation at real time rates. These characteristics permit the use of catheter-like US transducers [19]. Although they have limited FOV, the small size makes them more easily worn for extended periods and might enable use in online estimation of muscle kinematics for real time exoskeleton control.

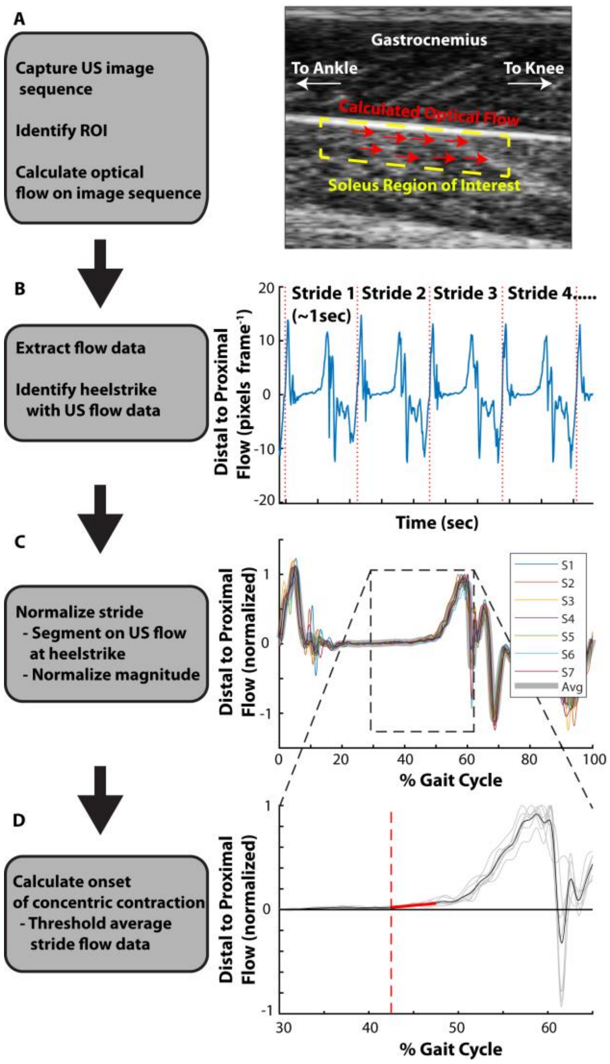

Figure 2: Processing steps to extract time of muscle concentric contraction. (A) We captured a sequence of US images with the person walking. Once captured, we identify the region of interest and the algorithm performs a sparse to dense optical flow routine to extract the flow parallel to the aponeurosis. **(B)** With the extracted flow data, the algorithm identifies the heel strike. **(C)** The flow data is normalized to a standard stride using the heel strike indices and peak magnitude is normalized to 1. **(D)** We isolate 30% - 55% phase of stride where concentric contraction occurs and threshold to determine the time when the muscle shortens against the tendon. Once the US images are captured, the algorithm runs at frequencies greater than 120Hz which is faster than the rate we can capture images.

III. Validation of Tracking Algorithm

Several experimental studies were conducted to validate the US measurement and to investigate the range of applications of this technology. All protocols were approved by the Harvard Medical School IRB and subjects gave written consent prior to participating.

A. Detection of Heel Strike

To evaluate the accuracy of using US data to detect heel strike, we calculated the error between when the US algorithm and the GRF determined heel strike (HS) [20]. Three healthy subjects walked on a treadmill at five imposed gait patterns which we thought would contribute to variability in gait (-5%, 0%, +5% cadence, weak and strong push-off). Muscle-based HS detection was on average 6.7 ± 7.7 ms later than the GRF-based HS detection for each participant (Fig 3). Given an average gait cycle of approximately 1000 ms for this walking speed, this difference was on average 0.67% of the gait cycle. This difference was within the sampling rate of the US system which is approximately 113Hz or 8 ms and represents less than

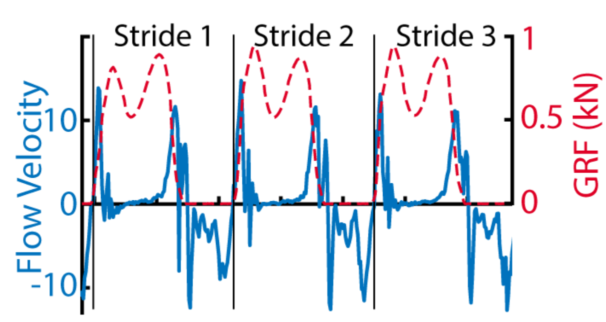

Figure 3: Representative comparison between optical flow (blue) and GRF (red). The vertical line is the US-determined start of the stride.

1% of the gait cycle. Due to known difficulties associated with walking dynamics of stroke survivors, we separately performed preliminary evaluation of foot-contact (heel-strike) detection on the paretic leg of stroke survivors [21]. The difference in the time of muscle-based and GRF-based HS detection was 0 ± 9 ms and 3 ± 7 ms for the two participants.

B. *Onset of Muscle Concentric Contraction*

To evaluate the ability of our automated process to detect onset of muscle concentric contraction, we performed a comparison to manually estimated onset which was similar to other studies that have used visual manual tracking [22, 23]. We performed the analysis for 16 walking samples that provide variability across individuals and variability in amount of muscle effort (7 individuals at 1.5 ms^{-1} free walking, 3 participants at 1.5 ms^{-1} for low, normal, high pushoff effort). A researcher visually observed frame sequences and estimated the frame number when proximal displacement began. Potential bias should be reduced as the frame number cannot be translated to percent gait cycle without also knowing the frame number of the heel-strike and performing the calculation. These frame numbers were recorded for 5-6 consecutive strides, converted to percent gait cycle using the frame number of onset and heel strike, and averaged for each condition. RMSE between automatic tracking and manual was 2.6% of the gait cycle (%GC). We then performed a linear regression to compare the automated and manual techniques (Fig 4). Automated detection of the onset of contraction followed similar trends to that of manually determined onset (R^2 =0.923; p <0.0001).

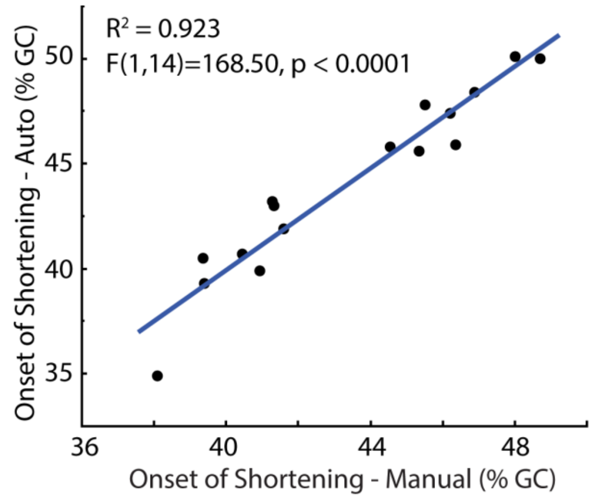

Figure 4: Comparison of manually and automatically determined onset

IV. Evaluation of Muscle Contraction patterns in Human Walking

A. *Effect of Individual Variability and Gait Pattern on Muscle Contraction Patterns*

To evaluate the system's ability to provide information about the onset of muscle concentric contraction which could be used to inform exosuit control, we examined the ability to measure inter-subject variability in onset of muscle concentric contraction, changes in onset timing due to changes in gait condition, and feasibility of detecting concentric contraction of paretic muscles in clinical (post-stroke) walking.

Individuality in onset of muscle concentric contraction: We expected to be able to detect differences in muscle timing across individuals. The onset of muscle contraction for the seven healthy subjects walking at 1.5 ms^{-1} ranged from 39.3%GC to 45.8%GC with an average of 42.8% ± 2.9%GC (Fig 5). Although individual data was not reported in prior studies, the transition to muscle concentric contraction shown from this technique aligns with reported group average data [13, 15].

Considering that the onset of positive joint velocity was fairly consistent across individuals, the determination that the onset muscle concentric contraction (*i.e.* onset of positive muscle power) is variable may help explain why there has been varied response of individuals to fixed exosuit assistive profiles in past studies [6, 24]. Perhaps coincidental but nevertheless interesting, the average onset time measured from these individuals is similar to what our group has previously measured as the best assistance onset time for group average metabolic improvement [2].

Effect of incline on onset of concentric contraction: The function of muscles as well as kinematics/kinetics adapt to different energetic requirements and environmental conditions [25-29]. Because changes in walking grade have a large impact on the joint energetic requirements and affect how well tendons can store and return energy, we evaluated the effect of incline on muscle contraction. We expected that muscle onset of concentric contraction would shift to an earlier time due to the necessity of adding addition energy. Four participants walked at 1.25 ms^{-1} at level and incline (10%) grade. Based off initial observations, we broadened the allowable region for detecting concentric contraction onset to begin at 20%GC. By changing from level walking to incline walking, we were able to detect a shift in contraction timing on average from 36.8 ± 4.8 %GC to 29.75 ± 5.7 %GC (Fig 6). The degree of shift in timing was on average 7 ± 5 %GC earlier in incline walking but was also variable and ranged from 14% earlier to 1% later. The findings supported our hypothesis that muscle onset of concentric contraction would change as individuals transition

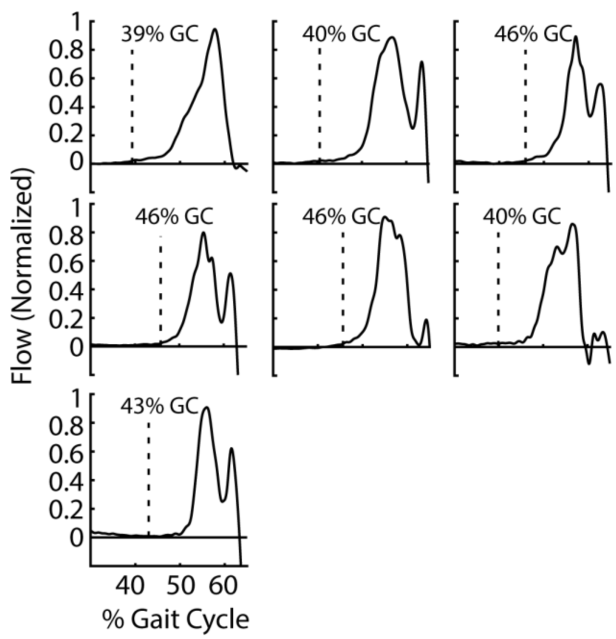

Figure 5: B-mode US flow data for 7 healthy participants. Vertical lines represent detected transition.

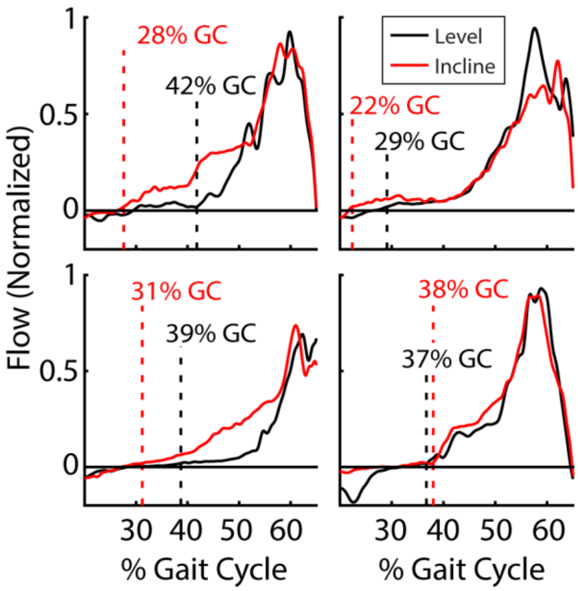

Figure 6: B-mode US flow data for 4 participants walking at level and incline (10%) grade. Vertical lines represent detected transition.

from level to incline walking. The impact of incline has been evaluated for the gastrocnemius [29], but this is the first time we are aware that it has been evaluated for the soleus. For exosuit control, this information could be used to update the assistance patterns as people walk in real-world conditions.

Feasibility in clinical populations: We also demonstrated the feasibility of estimating the time of muscle concentric contraction for stroke survivors. Two stroke survivors walked on a treadmill at their comfortable walking speed while we measured the muscle contraction on the paretic leg. The algorithm estimated an onset of contraction of 47.7% and 51% GC (Fig 7). We found that the signal to noise ratio for the flow data in the paretic muscle was higher for one of the individuals than for healthy. The magnitude of the flow vector was much smaller and suggest that the muscle contribution to power was likely smaller compared to healthy. These findings present interesting opportunities for use of muscle tracking in clinical populations. Our primary purpose for this work was the use of muscle contraction analysis for control of exosuits. Because stroke survivors cannot undergo prolonged optimization or parameter sweep protocols to determine optimal assistance profiles, this technique provides a potential approach for rapidly determining a useful profile. Furthermore, as stroke survivors improve, this approach could potentially allow for updating of assistance profiles as individuals improve. Beyond the use for exosuit control, this approach could be potentially used as a clinical tool for better understanding of the impaired muscle.

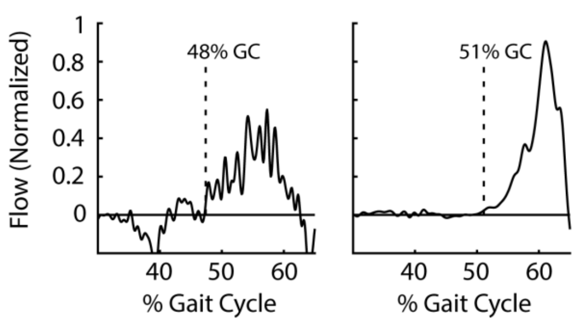

Figure 7: B-mode US flow data for 2 stroke participants. Vertical lines represent detected transition.

B. *Comparison of Ankle Joint and Muscle Dynamics*

Despite dynamic coupling between the muscle and joint through the AT, previous data has suggested that little relation can be drawn between muscle kinematics and joint kinematics [7]. We also evaluated this relationship with the individual data we collected to help determine if any relationships exist between the joint and onset of muscle concentric contraction.

The onset of muscle concentric contraction (*i.e.* positive muscle power) was, on average, 42.8 %GC and was significantly earlier (paired t-test; $p < 0.0001$) than the onset of positive ankle velocity (*i.e.* plantarflexion, positive power) at 49.9 %GC (Fig 8). The range in the difference was -2.4% to -13.5 %GC across individuals with an average of -6.6 ± 2.9 %GC. We found a stronger relationship between muscle onset time and the transition to positive ankle angular acceleration ($p = 0.0006$, $R^2 = 0.61$). The difference between muscle onset and acceleration zero crossing was -1.0 ± 3.2 %GC and we found no significant difference (paired t-test; $p = 0.3222$). The stronger relationship between the transition to positive angular acceleration and muscle onset in level ground walking suggests that the muscle may start concentrically contracting to begin the process of decelerating ankle dorsiflexion and beginning the process towards ankle push-off.

V. Preliminary Exosuit Response Evaluation

As part of a larger goal of improving exosuit performance, we evaluated whether muscle dynamics could be used to help develop individualized assistance profiles that are adaptive to changes in environmental demands. Efficient and powerful ankle plantarflexion is generated only through properly timed energy exchange between the muscle, the series elastic tendon,

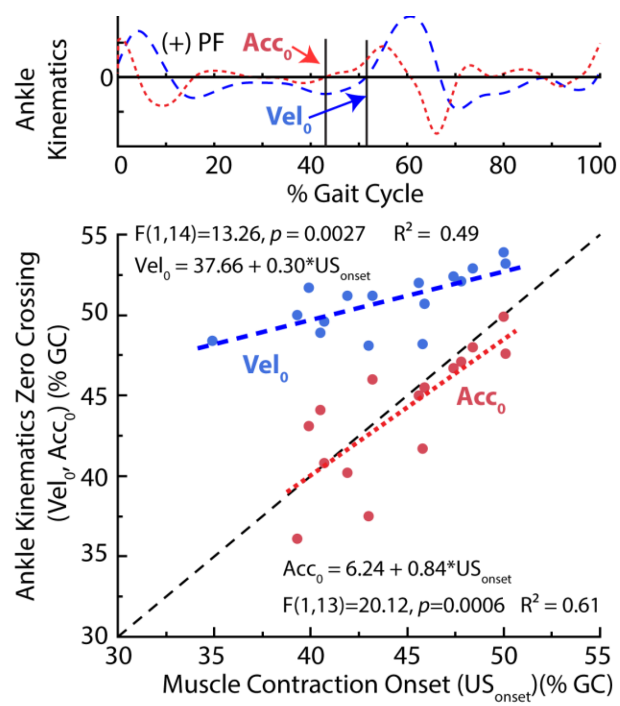

Figure 8: (Top) Representative plot of ankle angular velocity and acceleration. **(Bottom)** Relationship between onset of muscle concentric contraction and positive ankle velocity (blue) and acceleration (red).

the joint, body, and environment. Therefore, understanding and maintaining effective MTU dynamics is important for efficient walking [15, 30, 31]. In incline walking, because additional potential energy must be added to the center of mass, storage and return of energy in the muscle tendon-complex of the ankle is less effective [32]. Because the muscle is potentially adding this additional energy in incline walking, we expect be able to measure changes in onset of concentric contraction. We suspect that we can measure these mechanistic changes through onset of concentric contraction and use it to prescribe updated assistance profiles that deliver maximum metabolic benefit. We hypothesized that the onset of concentric contraction could be used to (1) predict an effective exosuit assistance time in baseline level walking and (2) predict what the assistance time should be updated to in incline walking. We expected that the updated assistance time for incline walking would outperform a fixed condition and the condition based on US time from level walking.

A. *Soft Exosuit and Preliminary Study Design*

We applied active ankle plantarflexion with a soft exosuit (Fig 9). The exosuit and its embedded sensors (IMUs and load cells) were similar to [33], but all components related to hip joints were removed to deliver the assistance purely to the wearer's ankle joints. An offboard actuation system [34] was used to generate assistive forces, and the forces were transmitted via Bowden cables to the exosuit. A MATLAB Simulink and Speedgoat (Mathworks, USA) platform was used to segment gait cycles and to control the actuators generating desired force profiles in real time. The spline-curved force profiles were designed to have a 300N peak at 57.5 %GC, while their onset timings could vary from 35 %GC to 47.5 %GC.

As a proof-of-concept, we collected data on a healthy male subject, who is experienced with walking with the exosuit. The protocol consisted of walking wearing the exosuit at 1.25 m/s on level and inclined (10 %) grade. For each grade, we tested the following assistance onset timings in GC: *Standard* (42.5 %GC), *Early* (35 %GC), *Late* (47.5 %GC), and *US*. The onset timing for *Standard* was derived from the group average assistance profile from our previous study that provided the best metabolic benefit [2]. As such, we expected this timing to provide good, if not the best, metabolic benefit. Note that for this subject, at level grade, the onset timing for *US* and *Standard* were the same (42.5 %GC). Early and late timing were chosen to cover a reasonable range of assistance timings. We also tested *Powered-off* for baseline comparison, where the system did not apply any force. For each condition, we measured whole-body energetics using a portable indirect calorimetry device (K5, COSMED, Italy) and a modified Brockway equation [35].

B. *Evaluation of Exosuit Performance*

In level grade walking, the participant's metabolic demand was reduced (-8.7%) for the *Standard* and *US-Level* assistance profile but was only marginally better than *Early* (Fig 10). In incline walking, the *Standard* and *US-Level* assistance time also performed well (-9%) and better than the *Early* (-8.2%) and *Late* (-7%). However, by updating the assistance time to the participants muscle onset time for incline, metabolic benefit was further improved to a 11% reduction. These preliminary results show that while the conditions for level ground can be partially generalized to incline walking, by updating the assistance profile to the user's biomechanical changes in muscle contraction, we may be able to improve performance of soft exosuits.

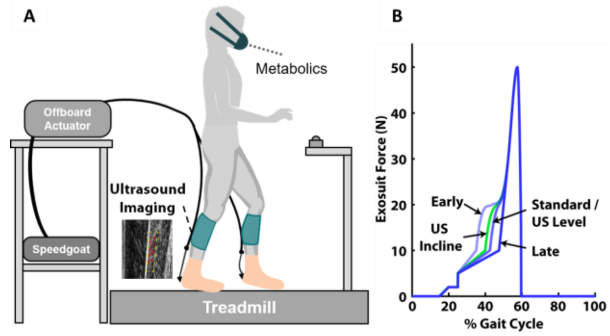

Figure 9: (A) Experimental setup and **(B)** assistance profiles

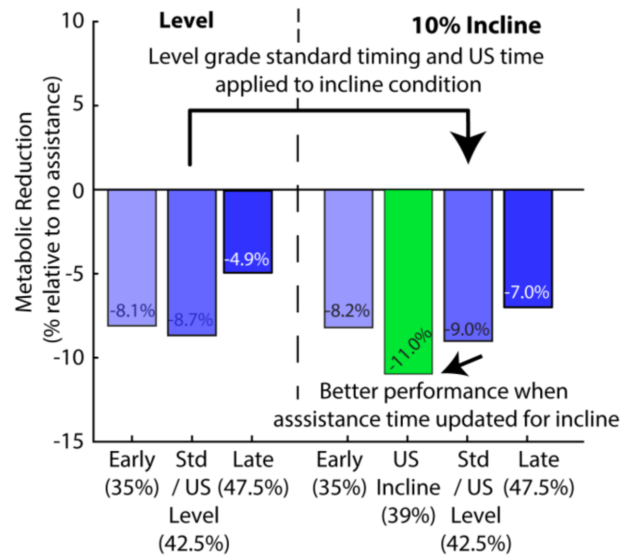

Figure 10: Metabolic reduction of one subject with different assistance time in level and 10% grade. Preliminary evaluation in use of US to update assistance times for an individual in variable gait conditions.

VI. Conclusion and Future Work

This paper presents an approach to automate the segmentation of gait and detection of the onset of muscle concentric contraction at real time rates using B-mode US. For wearable devices, a major benefit of the system is the small ROI which suggest that small catheter US transducers might be used. The technique detected individualized contraction times in healthy and stroke participants and changes in timing associated with incline walking. Finally, we provide initial proof-of-concept showing how the onset of muscle concentric contraction might be used for optimizing and updating assistance timing in variable gait conditions. Future work would need to be performed to evaluate if the trends hold across individuals, populations, and gait conditions.

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